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[54] LOW POWER X-RAY SOURCE WITH IMPLANTABLE PROBE FOR TREATMENT OF BRAIN TUMORS

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[*] Notice: The portion of the term of this patent subsequent to Oct. 16, 2009 has been disclaimed.

[21] Appl. No.: **955,494**

[22] Filed: **Oct. 2, 1992**

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Related U.S. Application Data

[63] Continuation-in-part of Ser. No. 577,883, Sep. 5, 1990, Pat. No. 5,153,900.

[51] Int. Cl.⁵ **H01J 35/14**

[52] U.S. Cl. **378/65; 378/64; 378/205**

[58] Field of Search 378/64, 65, 205, 208, 378/193, 195, 138, 137, 170; 606/33; 128/659; 600/1

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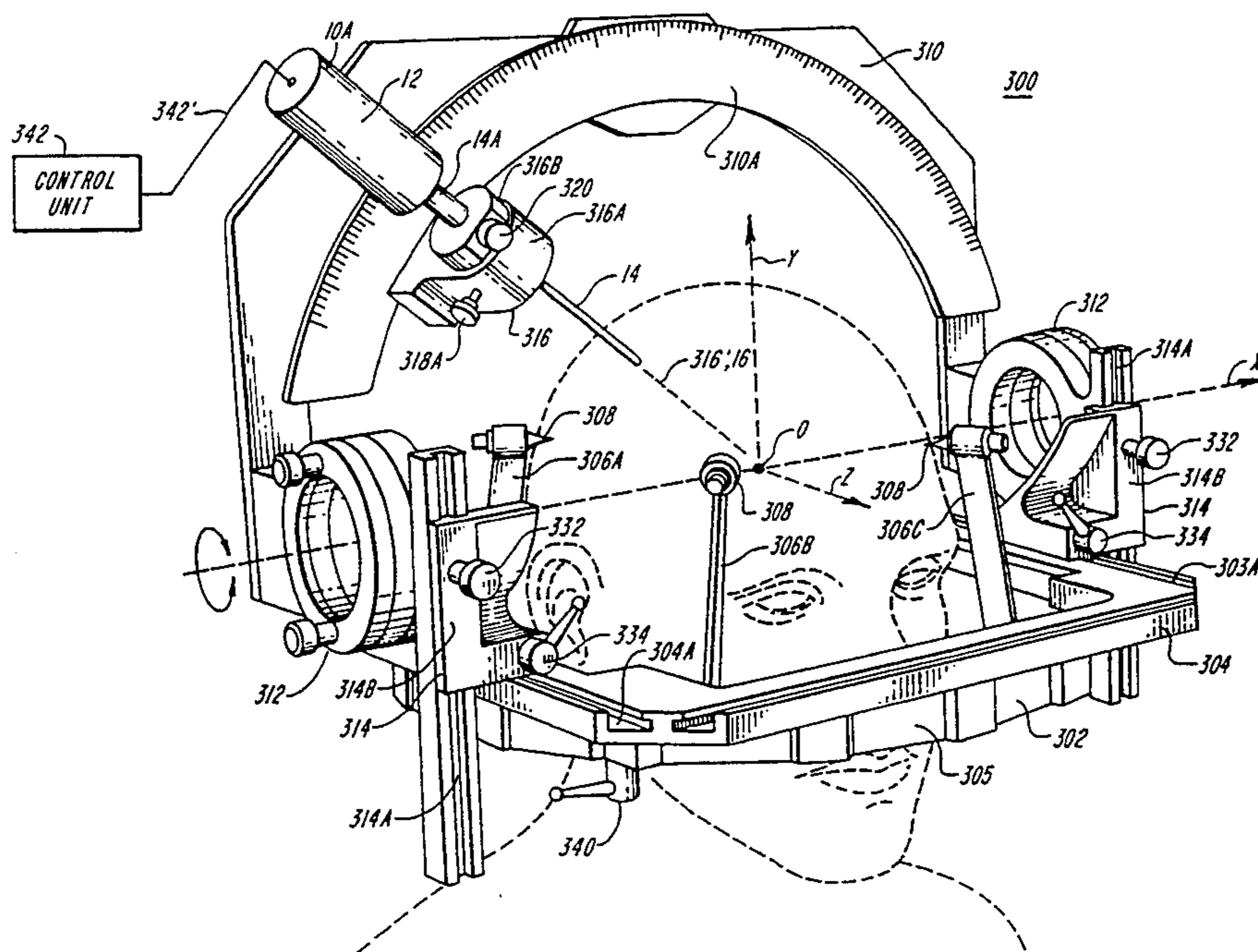
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Attorney, Agent, or Firm—Lappin & Kusmer

[57] ABSTRACT

An apparatus for treating a brain tumor in a patient, which includes an x-ray source of preselected or pre-programmed duration and intensity assembled in combination with a reference frame. The reference frame can be, for example, a stereotactic frame. The apparatus allows accurate positioning of the x-ray source within or adjacent to the desired region to be irradiated in the brain of the patient.

18 Claims, 9 Drawing Sheets



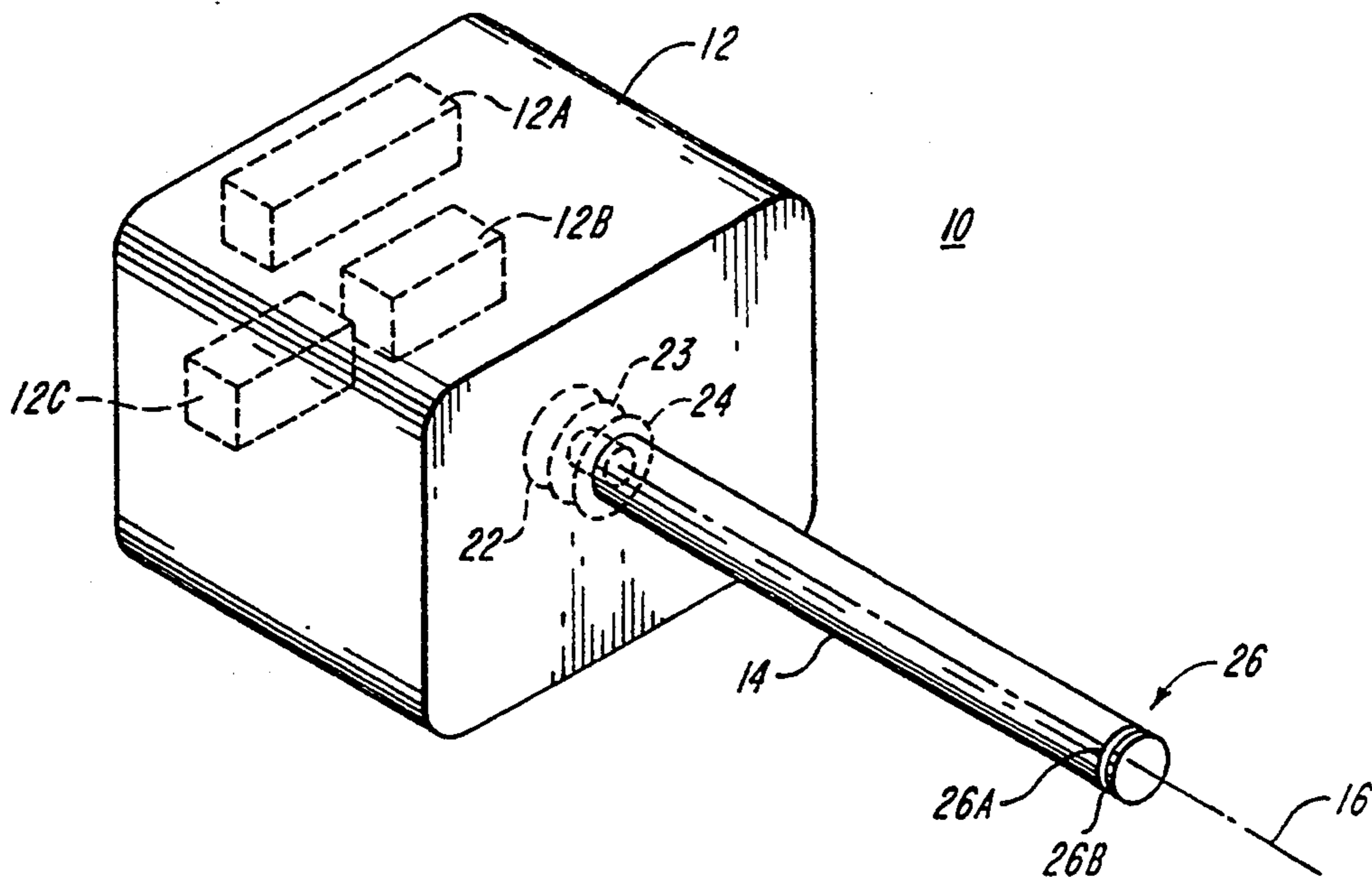


FIG. 1

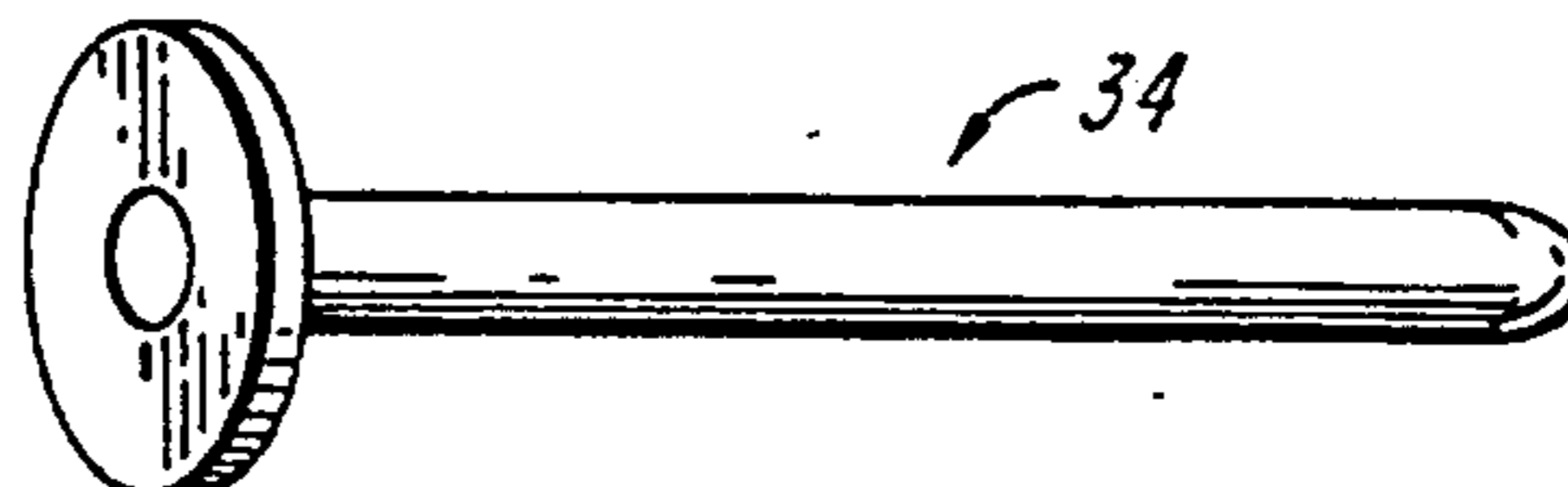


FIG. 2

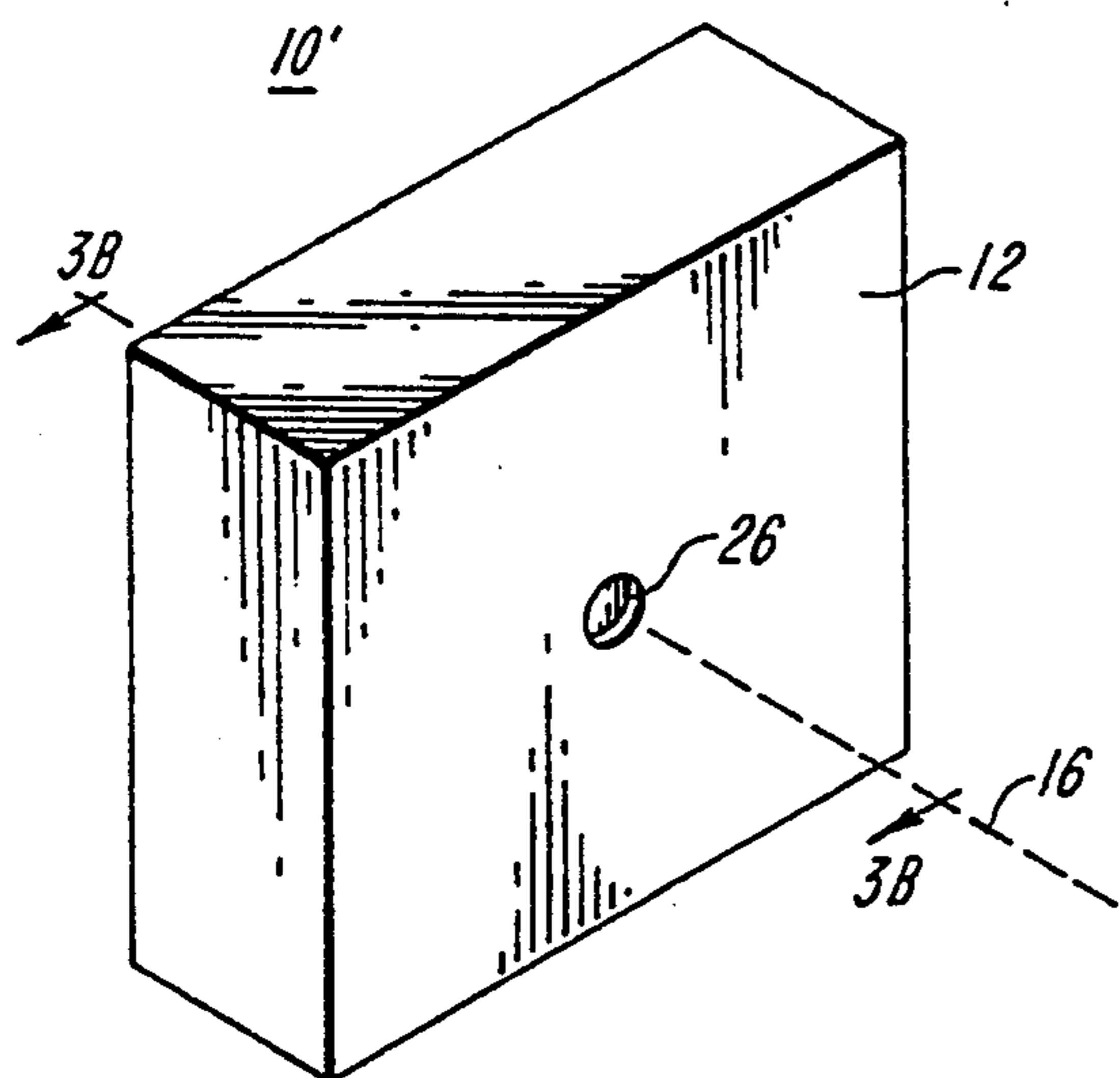


FIG. 3A

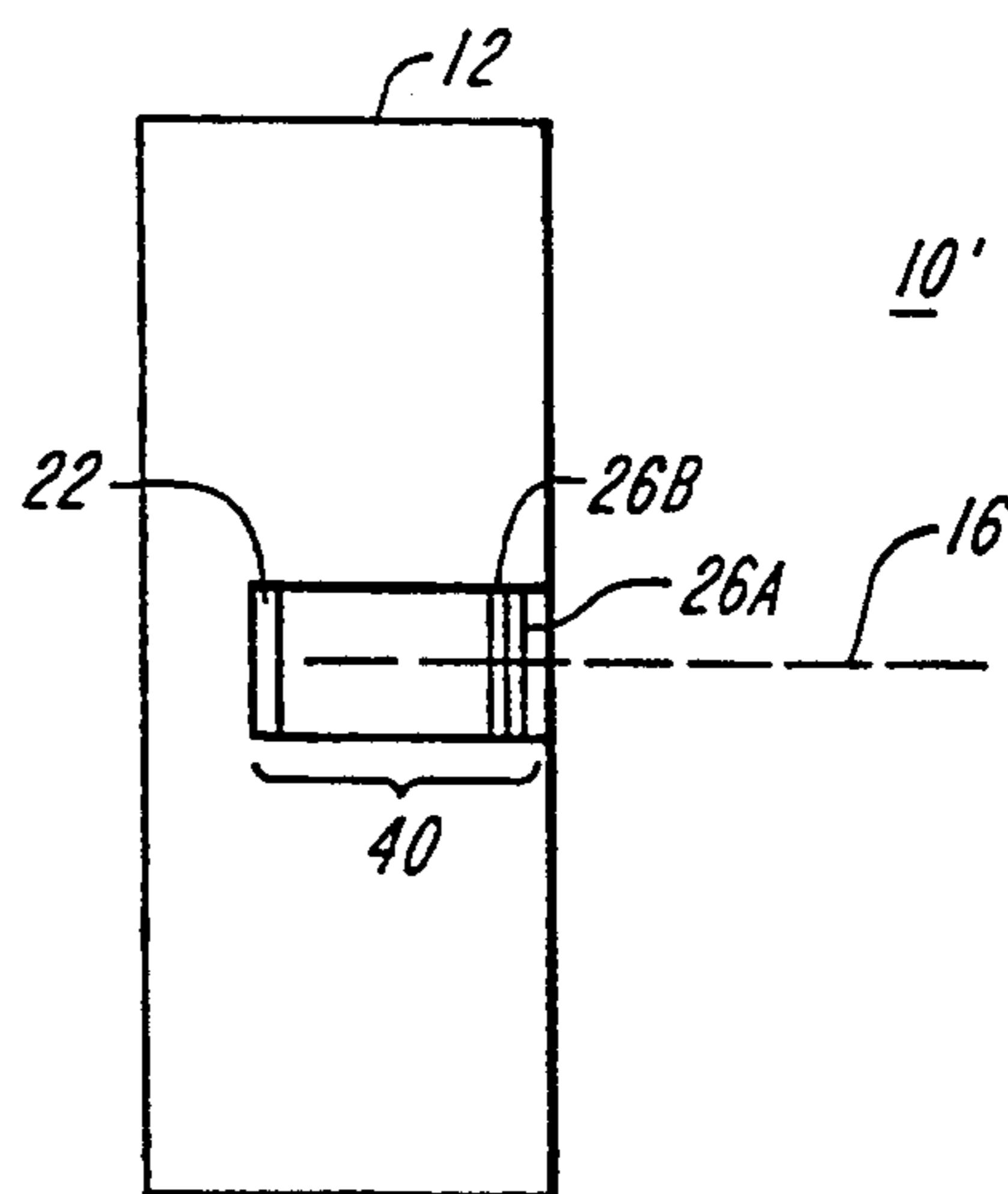


FIG. 3B

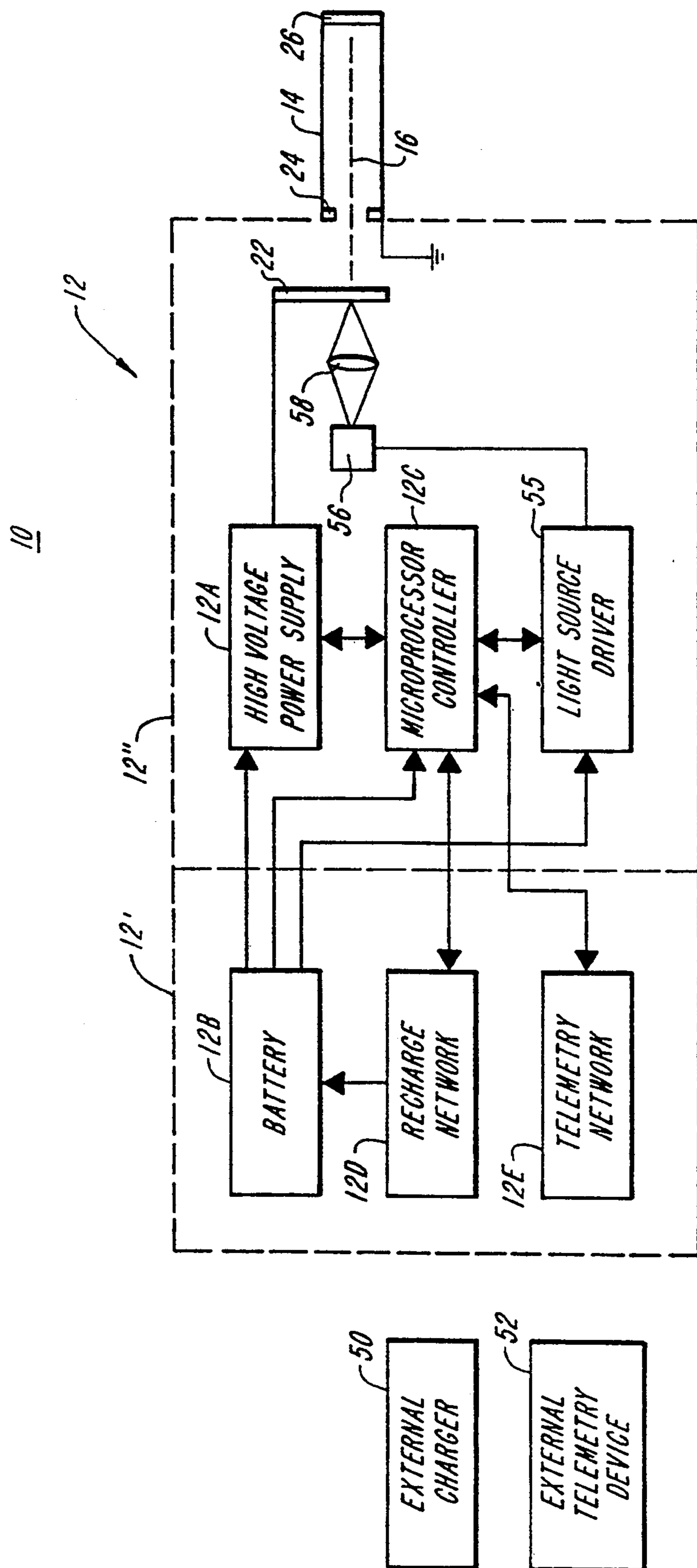


FIG. 4

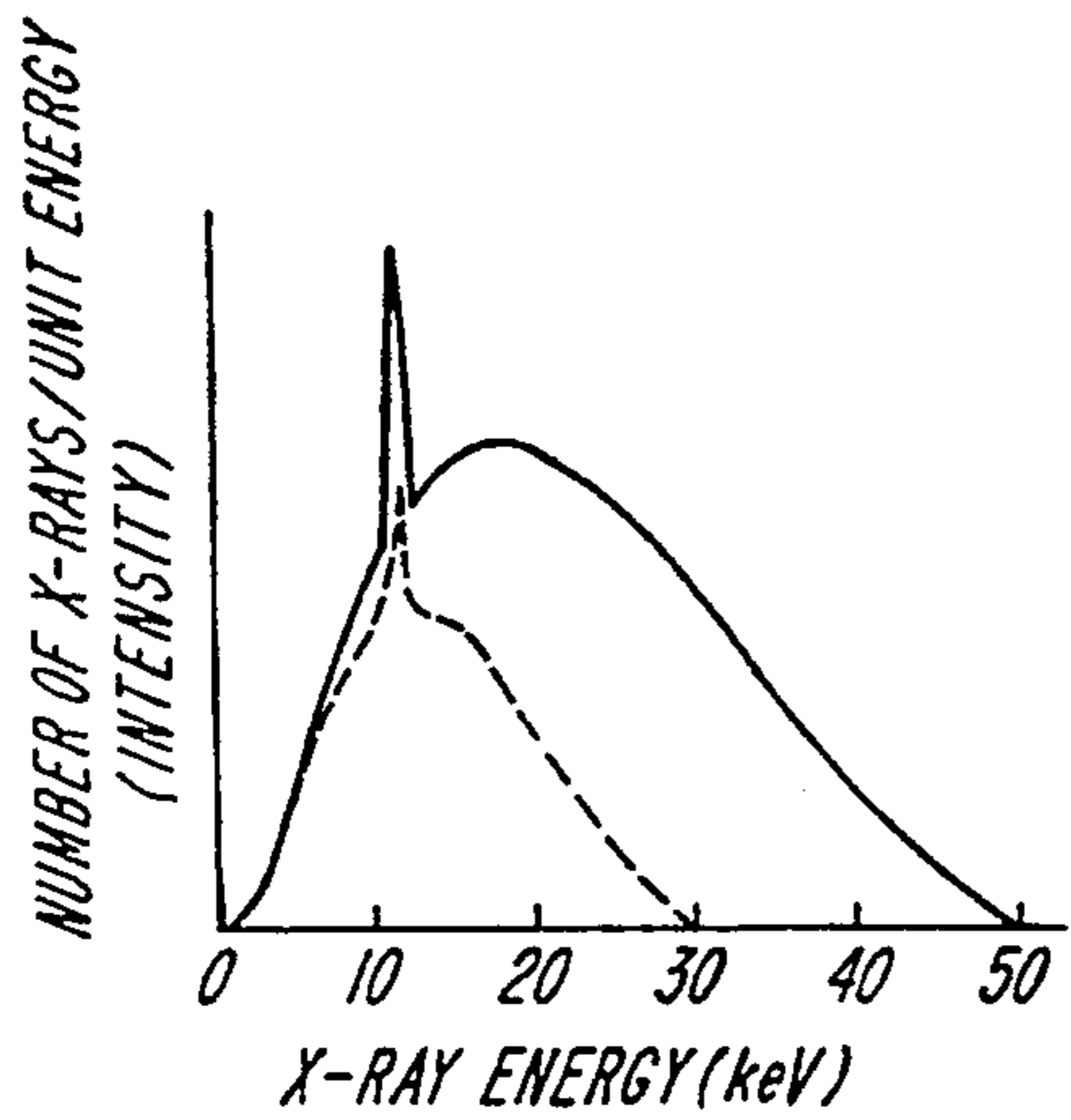


FIG. 5A

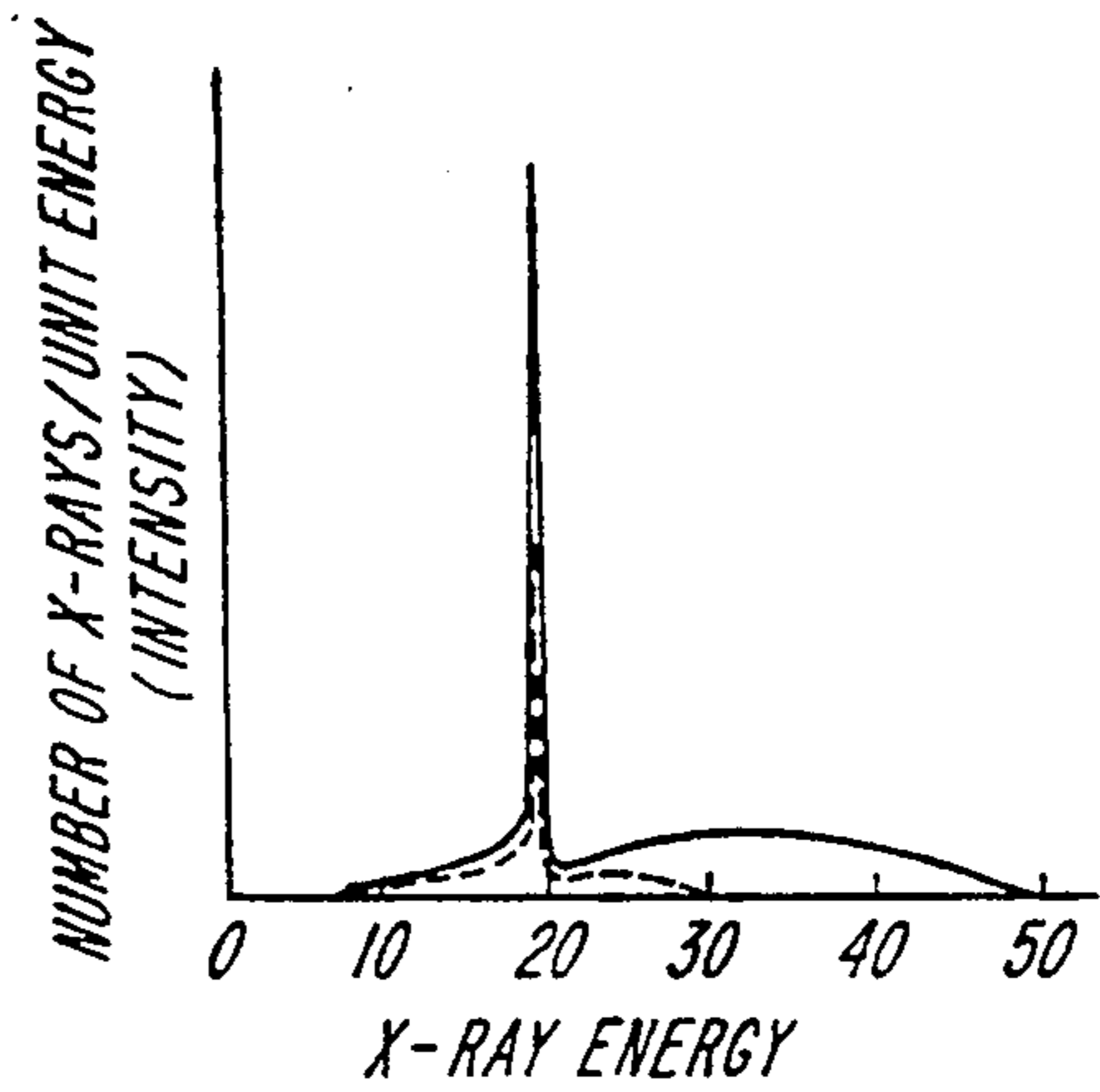


FIG. 5B

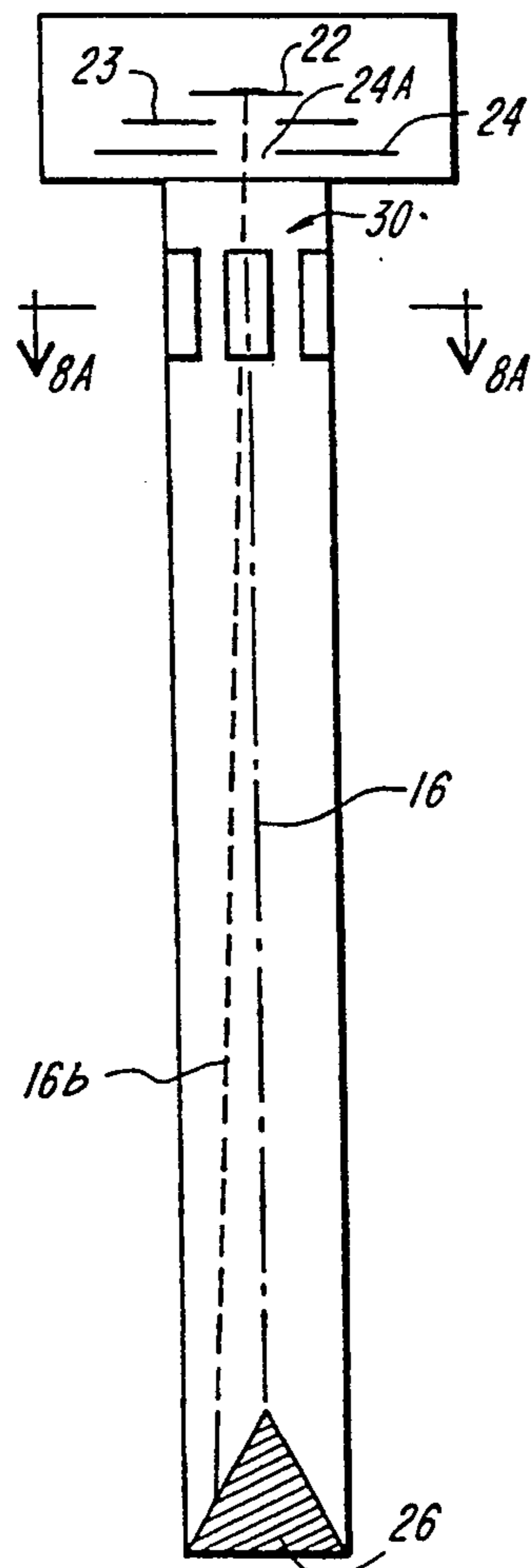


FIG. 8

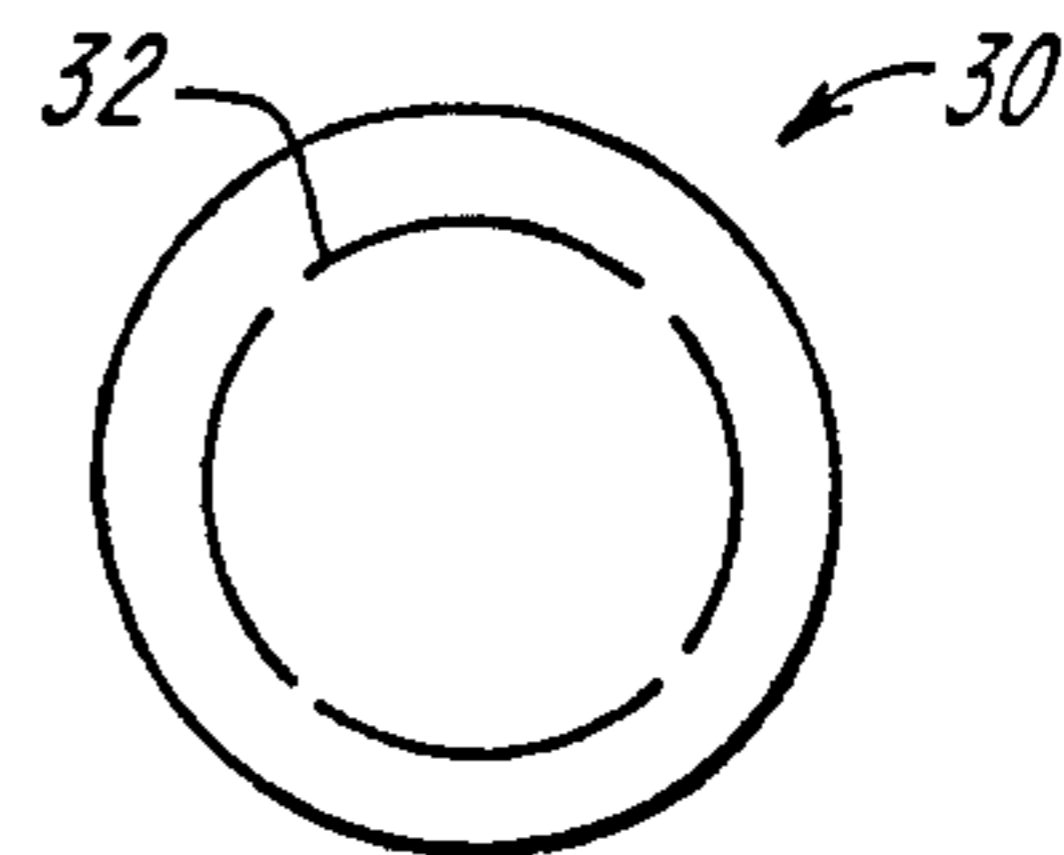


FIG. 8A

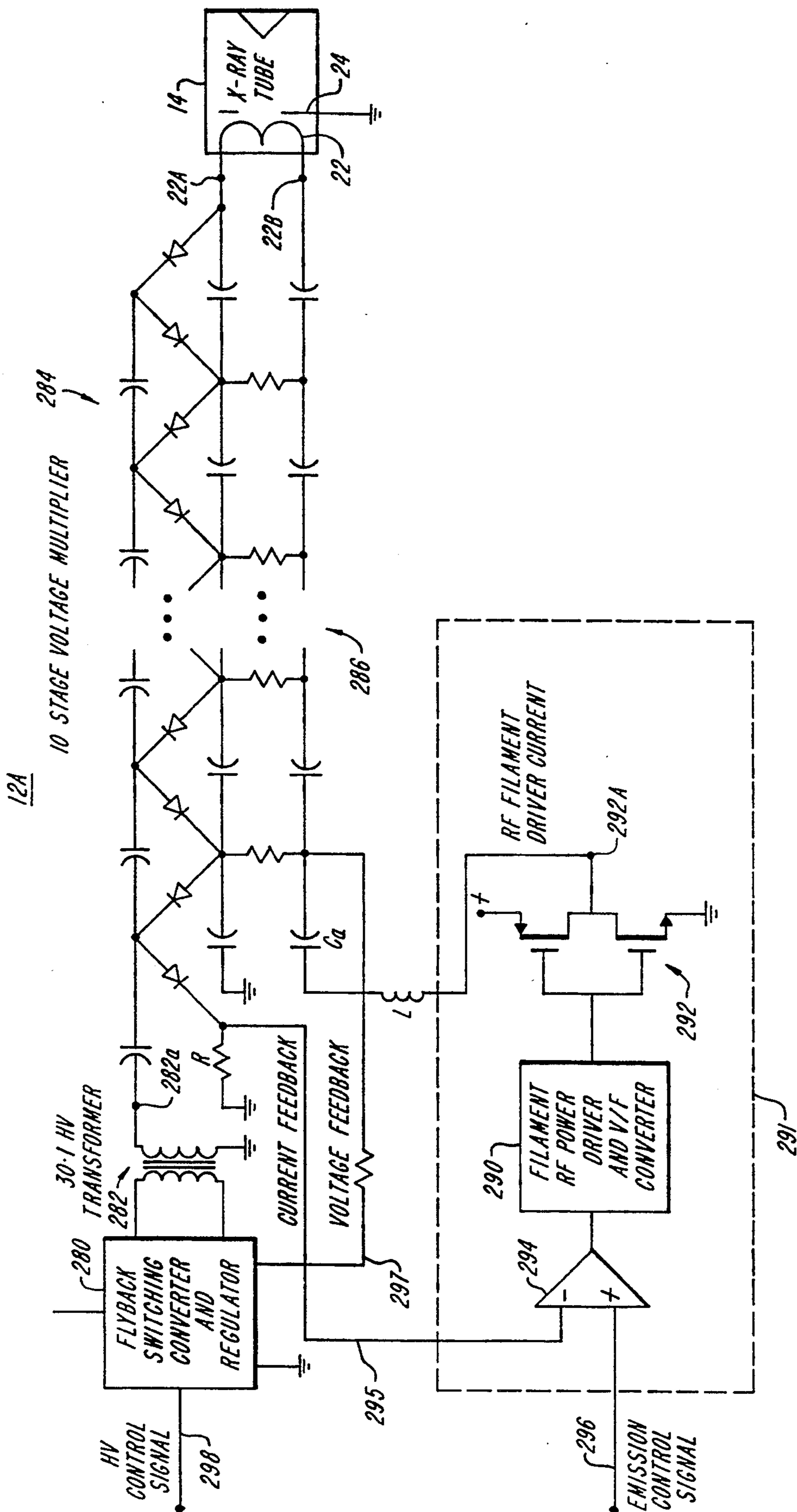


FIG. 6

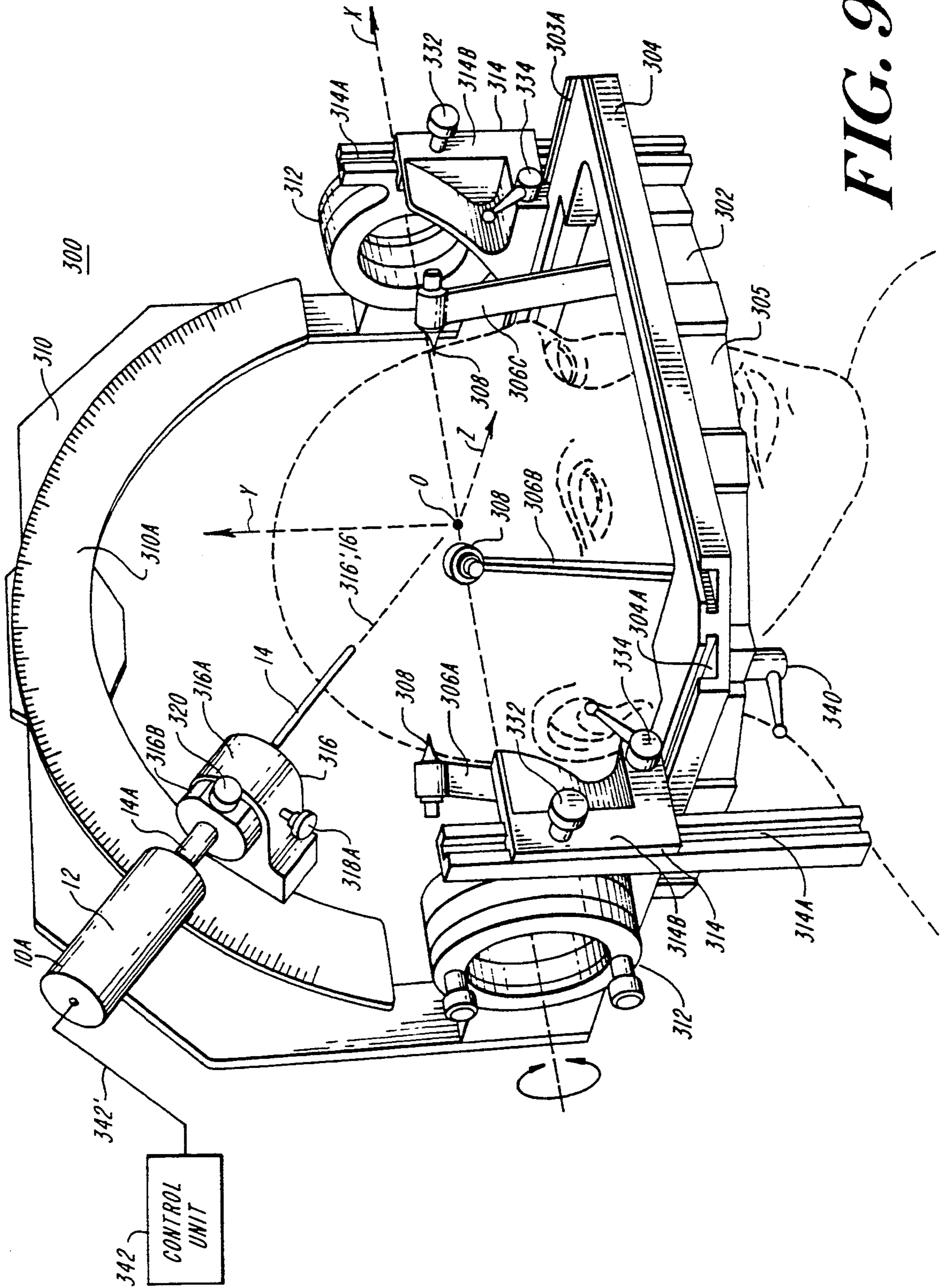


FIG. 9

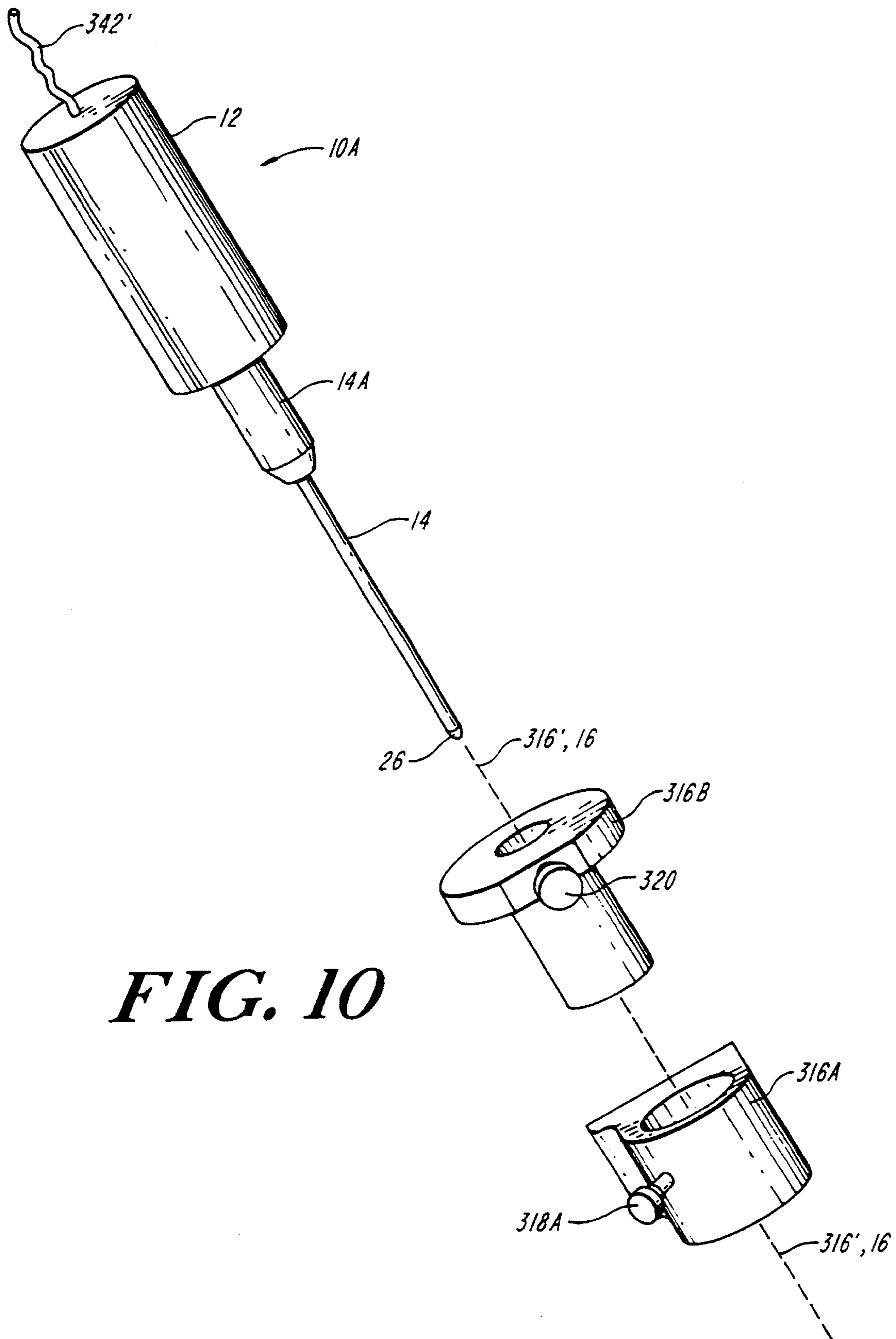


FIG. 10

10A

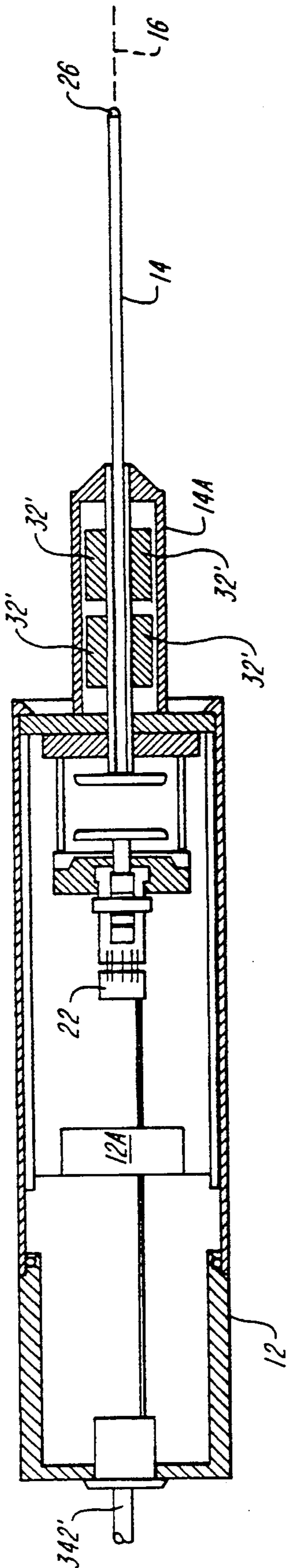


FIG. 11

LOW POWER X-RAY SOURCE WITH IMPLANTABLE PROBE FOR TREATMENT OF BRAIN TUMORS

REFERENCE TO RELATED APPLICATION

This application is a continuation-in-part of U.S. patent application Ser. No. 577,883, entitled "Miniaturized Low Power X-Ray Source," filed Sep. 5, 1990 (now U.S. Pat. No. 5,153,900), assigned to the assignee of the present application.

BACKGROUND OF DISCLOSURE

The present invention relates to a miniaturized, low power, programmable x-ray source for use in delivering low-levels of substantially constant or intermittent x-rays to a specified region.

Conventional medical x-ray sources are large, fixed position machines. Generally, the head of the x-ray tube is placed in one room and the control console in an adjoining area, with a protective wall, equipped with a viewing window, separating the two. The x-ray tube typically is approximately 20 to 35 centimeters (cm) long, and approximately 15 cm in diameter. A high voltage power supply is housed within a container located in a corner of the room containing the x-ray tube. Patients are brought to the machine for diagnostic, therapeutic, or palliative treatment.

Diagnostic x-ray machines are typically operated at voltages below 150 kilovolts (kV), and at currents from approximately 25 to 1200 milliamps (mA). By contrast, the currents in therapeutic units typically do not exceed 20 mA at voltages which may range above 150 kV. When an x-ray machine is operated at nominal voltages of 10 to 140 kV, the emitted x-rays provide limited penetration of tissue, and are thus useful in treating skin lesions. At higher voltages (approximately 250 kV), deep x-ray penetration is achieved, which is useful in the treatment of major body tumors. Supervoltage machines, operable in the 4 to 8 megavolt (MV) region, are used to ablate or destroy all types of tumors, except superficial skin lesions.

A conventional x-ray tube includes an anode, grid, and cathode assembly. The cathode assembly generates an electron beam which is directed to a target, by an electric field established by the anode and grid. The target in turn emits x-ray radiation in response to the incident electron beam. The radiation absorbed by a patient generally is that which is transmitted from the target in the x-ray tube through a window in the tube, taking into account transmission losses. This window typically is a thin section of beryllium, or other suitable material. In a typical x-ray machine, the cathode assembly consists of a thoriated tungsten coil approximately 2 mm in diameter and 1 to 2 cm in length which, when resistively heated with a current of 4 amps (A) or higher, thermionically emits electrons. This coil is surrounded by a metal focussing cup which concentrates the beam of electrons to a small spot on an opposing anode which also functions as the target. In models having a grid, it is the grid which both controls the path of the electron beam and focuses the beam.

The transmission of an electron beam from cathode to anode is influenced by electron space charge forces which tend to become significant in conventional x-ray machines at currents exceeding 1 A. In such conventional machines, the beam is focussed on the anode to a spot diameter ranging anywhere from 0.3 to 2.5 milli-

meters (mm). In many applications, most of the energy from the electron beam is converted into heat at the anode. To accommodate such heating, high power medical x-ray sources often utilize liquid cooling and a rapidly rotating anode, thereby establishing an increased effective target area, permitting a small focal spot while minimizing the effects of localized heating. To achieve good thermal conductivity and effective heat dissipation, the anode typically is fabricated from copper. In addition, the area of the anode onto which an electron beam is incident requires a material of high atomic number for efficient x-ray generation. To meet the requirements of thermal conductivity, effective heat dissipation, and efficient x-ray generation, a tungsten alloy typically is embedded in the copper.

In use, the total exposure from an x-ray source is directly proportional to the time integral of the electron beam. During relatively long exposures (e.g. lasting 1 to 3 seconds), the anode temperature may rise sufficiently to cause it to glow brightly, accompanied by localized surface melting and pitting which degrades the radiation output. However, thermal vaporization of the tube's coiled cathode filament is most frequently responsible for conventional tube failure.

While the efficiency of x-ray generation is independent of the electron beam current, it is highly dependent on the acceleration voltage. Below 60 kV, only a few tenths of one percent of the kinetic energy from an electron is converted to x-rays, whereas at 20 MV that conversion factor rises to 70 percent. An emitted x-ray spectrum is composed in part of discrete energies characteristic of transitions between bound electron energy levels of the target element. The spectrum also includes an x-ray energy continuum, known as bremsstrahlung, which is caused by acceleration of the beam electrons as they pass near target nuclei. The maximum energy of an x-ray cannot exceed the peak energy of an electron in the beam. Further, the peak of the bremsstrahlung emission curve occurs at approximately one-third the electron energy.

Increasing the electron current results in a directly proportional increase in x-ray emission at all energies. However, a change in beam voltage results in a total x-ray output variation approximately equal to the square of the voltage, with a corresponding shift in peak x-ray photon energy. The efficiency of bremsstrahlung radiation production increases with the atomic number of the target element. The peak output in the bremsstrahlung curve and the characteristic spectral lines shift to higher energies as the atomic number of the target increases. Although tungsten ($Z=74$) is the most common target material used in modern tubes, gold ($Z=79$) and molybdenum ($Z=42$) are used in some specialty tubes.

X-rays interact in several ways with matter. For biological samples, the following two types of interactions are most important: Compton scattering of moderate-energy x-rays with outer shell electrons; and, photoionizing interactions of inner shell electrons. In these processes, the probability of atom ionization decreases with increasing photon energy in both soft tissue and bone. For the photoelectric effect, this relationship follows an inverse third-power law.

One disadvantage of present x-ray devices used for therapy is the high voltage required when directed to soft tissue within or beneath bone. One example is in directing x-rays to areas of the human brain, which is

surrounded by bone. High energy x-rays are required to penetrate the bone, but often damage the skin and brain tissue. Another example in radiation therapy is in directing the x-rays to soft tissue located within the body cavity, couched among other soft tissue, or within an internal calciferous structure. Present high-voltage machines are limited in their ability to selectively provide desired x-ray radiation to such areas.

Another disadvantage of the high voltage output of present x-ray sources is the damage caused to skin external to the affected organ or tissue. Therefore, high voltage devices of present systems often cause significant damage not only to the target region or tissue, but also to all surrounding tissue and surface skin, particularly when used for human tumor therapy. However, since present devices apply x-ray radiation to target regions internal to a patient from a source external to the target region, such incidental tissue damage is practically unavoidable.

Specifically as to brain tissue, which lacks any substantial regenerative ability, the treatment of brain tumors requires precise techniques to bring about specific tissue destruction. The use of conventional x-ray devices in brain tumor therapy often lacks the precision needed in volumetric irradiation, resulting in the damage of non-cancerous tissue of the brain and associated glandular structures.

An alternative form of tumor therapy, called brachytherapy, involves implanting encapsulated radioisotopes in or near the tumor to be treated. While such use of radioisotopes may be effective in treating certain types of tumors, introduction of the isotopes requires invasive procedures which have potential side-effects, such as the possibility of infection. Moreover, brain swelling may occur in some applications because the emission from the isotope cannot be controlled. Further, there is no ability to provide selective control of time dosage or radiation intensity. Handling and disposal of such radioisotopes involves hazards to both the individual handler and the environment.

Invasive techniques of the brain require precise control of irradiation through the choice and concentration of isotopes used. Intracranial penetration poses a significant risk as is well known in the art.

In view of the above requirements and limitations to the use of x-rays from present machines in therapeutic, diagnostic, palliative, or evaluative environments, there remains a need for a relatively small, easily manipulated, low-energy, x-ray device. Such a device operating at low energy and power will be suitable for many of the applications described herein.

Thus, it is an object of the present invention to provide an easily manipulated, low-power x-ray device.

It is another object of the invention to provide a relatively small, low-power x-ray device having a controllable, or programmable, power supply.

It is another object of the invention to provide a relatively small, low-power x-ray device which is implantable into a patient for directly irradiating a desired region of tissue with x-rays.

It is yet another object of the invention to provide a relatively small, surface-mountable, low-power x-ray device for affecting a desired surface region with x-rays.

It is yet another object of the invention to provide a relatively small, low-power x-ray device which is partially implantable into a patient for directly irradiating a specified region with x-rays.

It is yet another object of the invention to provide a small, low-power x-ray device and reference frame assembly for controllably positioning an x-ray source within a patient's skull in order to irradiate and therefore treat a brain tumor.

SUMMARY OF THE INVENTION

Briefly, the invention is an easily manipulated apparatus having a low-level, electron beam (e-beam) activated x-ray source of preselected, or adjustable, duration, effective energy and intensity. In medical applications, the apparatus may be fully or partially implanted into, or surface-mounted onto a desired area of a patient to irradiate a preselected region with x-rays. In particular, the apparatus can be assembled in combination with a reference frame, for example, a stereotactic frame, and an associated coupler for use in the treatment of brain tumors.

The apparatus operates at a relatively low voltage, for example, in the range of approximately 10 kV to 90 kV, with small electron currents, for example, in the range of from approximately 1 nA to 100 μ A. To achieve a desired radiation pattern over a desired region, while minimally irradiating other regions, x-rays may be emitted from a nominal, or effective "point" source located within or adjacent to the desired region-to-be-irradiated. In some applications, a low dose rate of x-rays irradiates any part of the desired region, either continually or periodically, over extended periods of time. For use with a reference frame for treatment of brain tumors, a high dose rate for single dose irradiation is generally preferred.

The apparatus includes a controllable, or programmable, power supply located outside the desired region-to-be-irradiated to enable variations in voltage, current, and timing of an electron beam. The electron beam is controlled to pass along a desired beam axis and to be incident on a target which is preferably located in the patient's body. The axis may be straight, or curved. The composition and/or geometry of the target, or x-ray emitting, material is selected to provide a customized pattern of x-rays. Shielding at the emission site, or around the target, further enables control of the energy and spatial profile of the x-ray emission to match the preselected distribution of radiation throughout the desired region.

The present invention further provides a method of treating malignant cells, such as found in tumors, in vivo, utilizing the apparatus described above. Generally, the method involves identifying and locating malignant cells with a device generally available in the art, such as by computed tomography (CT) scanning or magnetic resonance imaging (MRI). A needle-type biopsy of the tumor may be performed to confirm the diagnosis. Then the region of treatment is selected and the radiation dosage determined. Such radiation treatment planning involves defining the size and shape of the tumor determining precisely its location in the body, identifying radiation-sensitive critical biological structures surrounding the tumor, deciding on the proper radiation dose distribution in the tumor and surrounding tissue and the entry path in to the tumor of the implanted portions of the apparatus. For spherical tumors, treatment planning can be performed manually using CT or MRI data. However, for more complex geometries, close-by critical structures, or higher precision procedures, computer-based "3-D" imagery is performed. In that case, tumors and critical structures are,

for example, manually or semiautomatically segmented on a series of digitized CT scans, and a 3-D composite is rendered, which allows viewing the tumor from any direction. Various software systems have been developed for radiosurgical procedures, such as the linac and gamma knife, and some are commercially available. For example, Radionics Software Applications of Brookline, Mass. offers for sale software which images the CRW and BRW stereotactic frame affixed to a graphically transparent skull. Isodose profiles are overlaid on the tumor and other brain tissue. Similar, software can be developed that allows imaging the present invention, affixed to the stereotactic frame, with the X-ray-radiating electron beam target imbedded in the tumor. Isodose contours around the target would be superimposed on the tumor and adjacent tissue. The absolute radiation dosage delivered along each contour would be determined by experimental dosimetry performed to calibrate the probe. In these tests, the dose is measured at multiple locations around the target immersed in a biological tissue-simulating phantom. Such plastic, "solid water," phantoms are commercially available (RMI, Middleton, Wis.) and simulate various body tissues, e.g., soft tissue of the brain. Either thermoluminescent detectors (TLD) or calibrated X-ray sensitive film (e.g., gafchromic film from Far West Technologies, Goleta, Calif.) can be positioned in the solid water to measure the dose directly. Using the imaging and dosimetry results from the radiation treatment planning, a low-power electron beam source and a selectively shaped x-ray radiation pattern generating target and shield assembly are positioned within or proximal to a region containing the malignant cells. The target and shield assembly geometry and materials are shaped and selected in accordance with the characteristics of the target volume. A programmable power supply is provided, which may be used to vary the voltage, current, and duration of the electron beam source to establish, in accordance with dosimetry information, a desired electron beam which is directed to the target. Finally, x-rays emitted from the target and shield assembly are transmitted to the malignant cells for selective destruction of the cells.

In particular, the treatment of a brain tumor can be carried out utilizing an apparatus of the present invention comprising the combination of a low-power x-ray source for generation of a controllable irradiation pattern, with a device for accurately positioning the x-ray source in the brain. The x-ray source can thus be precisely located near or in the tumor.

BRIEF DESCRIPTION OF DRAWINGS

The foregoing and other objects of this invention, the various features thereof, as well as the invention itself, may be more fully understood from the following description, when read together with the accompanying drawings in which:

FIG. 1 is a perspective view of a low power x-ray source embodying the present invention;

FIG. 2 is a schematic representation of a sheath adapted for use with the apparatus of FIG. 1;

FIGS. 3A and 3B are a perspective view and sectional view, respectively, of a surface-mountable apparatus embodying the present invention;

FIG. 4 is a schematic block diagram of the embodiment of FIG. 1;

FIGS. 5A and 5B are graphical representations of the x-ray emission spectrum of tungsten- and molybdenum-targets, respectively;

FIG. 6 is a detailed block diagram of the representative power supply of the embodiment of FIG. 1;

FIG. 7 is a detailed schematic diagram of the representative power supply of the embodiment of FIG. 1;

FIG. 8 is a perspective view of a beam steering assembly embodying the present invention;

FIG. 8A is a cross-section view of the assembly of FIG. 8, taken along lines 8a;

FIG. 9 is a perspective view of a brain tumor x-ray treatment system;

FIG. 10 is an exploded perspective view of an x-ray apparatus and the coupling assembly of the system of FIG. 9;

FIG. 11 is a cross-section view of the x-ray device of the system of FIG. 9; and

FIG. 12 is a schematic representation of the high voltage power supply of the x-ray device of the system of FIG. 9.

Like numbered elements in each FIGURE represent the same or similar elements.

DESCRIPTION OF THE PREFERRED EMBODIMENTS

The present invention is a relatively small, electron-beam activated, low power x-ray apparatus. The apparatus may be used for medical purposes, for example, therapeutic or palliative radiation treatment of tumors, or for other purposes.

With particular regard to medical uses, the apparatus may be fully implanted or partially inserted into a preselected internal region of a patient to provide x-ray radiation over selected exposure times. Alternately, the apparatus may be mounted on a surface of a patient external to a region to be irradiated. Also disclosed is a method for treating tumors in a patient, using the apparatus of the invention.

Generally, the apparatus of the present invention includes an electron-beam (e beam) activated x-ray source which operates at relatively low voltages, i.e. in the range of approximately 10 kV to 90 kV, and relatively small electron beam currents, i.e. in the range of approximately 1 nA to 100 μ A. At those operating voltages and currents, the x-ray output is relatively low, and the apparatus may be made quite small and be adapted for implantation in medical therapeutic applications. In view of the low level x-ray output, adequate tissue penetration and cumulative dosage may be attained by locating the x-ray source adjacent to or within the region to be irradiated. Thus, the x-rays are emitted from a well-defined, small source located within or adjacent to the region to be irradiated. In the preferred embodiment, a low dose rate of x-rays may be applied to any part of a tumor, either continually or periodically, over extended periods of time, e.g., up to one month. In use with a stereotactic frame for the treatment of brain tumors, a high dose rate may be applied to a tumor for shorter periods of time (i.e., on the order of 5 minutes to 3 hours).

The present invention provides interstitial radiotherapy similar to that achieved with implanted capsules, needles, tubes, and threads containing natural or artificial radioactive isotopes, known as brachytherapy. However, a programmable power supply may be included in the x-ray source of the present apparatus to vary energy, intensity, and duration of the radiation. This dif-

fers from brachytherapy in that the intensity and penetration depth of the x-rays may be changed without surgically or invasively replacing the isotopes. Furthermore, the present invention is not limited by the half-life of a particular isotope, and does not pose a radiation hazard when turned off.

FIG. 1 shows an x-ray apparatus 10 embodying the present invention. Apparatus 10 includes a housing 12 and an elongated cylindrical probe 14 extending from housing 12 along a reference axis 16. The housing 12 encloses a high voltage power supply 12A (illustrated in electrical schematic form in FIGS. 6 and 7). The probe 14 is a hollow tube having an electron beam generator (cathode) 22 adjacent to the high voltage power supply 12A. Cathode 22 is located in close proximity to an annular focussing electrode 23 typically at the same potential as the cathode 22. An annular anode 24 is positioned approximately 0.5 cm or more from the annular focussing electrode 23. A hollow, tubular probe 14 extends along the same axis as the cathode, grid, and the hole in the anode. Probe 14 is integral with the housing 12 and extends toward a target assembly 26. In various embodiments, parts of the probe 14 may be selectively shielded to control the spatial distribution of x-rays. In addition, the probe 14 may be magnetically shielded to prevent external magnetic fields from deflecting the beam away from the target.

The electron beam generator 22 may include a thermionic emitter (driven by a floating low voltage power supply) or a photocathode (irradiated by an LED or laser source). The high voltage power supply establishes an acceleration potential difference between the cathode of generator 22 and the grounded anode 24 so that an electron beam is established along the reference axis 16, through the center hole of the anode and to the target assembly 26, with the region between anode 24 and the target assembly 26 being substantially field free. The beam generation and acceleration components are adapted to establish a thin (e.g. 1 mm or less in diameter) electron beam within the probe 14 along a nominally straight axis 16.

In a preferred embodiment, the probe 14 is a hollow, evacuated beryllium (Be), Mo—Re or mu metal cylinder 15 cm long, with an interior diameter of 2 mm, and an exterior diameter of 3 mm. The target assembly 26 includes an emission element consisting of a small beryllium (Be) window element 26a coated on the side exposed to the incident electron beam with a thin film or layer 26b of a high-Z element, such as tungsten (W) or gold (Au). By way of example, with electrons accelerated to 30 keV-, a 2.2 micron thick tungsten film absorbs substantially all the incident electrons, while transmitting approximately 95% of any 30 keV-, 88% of any 20 keV-, and 83% of any 10 keV- x-rays generated in that layer. In the preferred embodiment, the beryllium substrate is 0.5 mm thick with the result that 95% of these x-rays generated in directions normal and toward the substrate, and having passed through the tungsten target, are then transmitted through the beryllium substrate and outward at the distal end of probe 14. While the window element 26a shown in FIG. 3b is in the form of a disc, other shaped elements may be used, such as those having hemispherical or conical outer surfaces.

In some forms of the invention, the window element 26a assembly may include a multiple layer film 26b, where the differing layers may have different emission characteristics. By way of example, the first layer may have an emission (vs. energy) peak at a relatively low

energy, and the second (underlying) layer may have an emission (vs. energy) peak at a relatively high energy. With this form of the invention, a low energy electron beam may be used to generate x-rays in the first layer (to achieve a first radiation characteristic) and high energy electrons may be used to penetrate through to the underlying layer (to achieve a second radiation characteristic). As an example, a 0.5 mm wide electron beam is emitted at the cathode and accelerated to 30 keV- through the anode, with 0.1 eV transverse electron energies, and arrives at the target 26 sixteen centimeters downstream from the anode, with a beam diameter of less than 1 mm at the target. X-rays are generated in the target material in accordance with preselected beam voltage, current, and target composition. The x-rays thus generated pass through the beryllium window 26a in the probe with minimized loss in energy. The window 26a may be made of carbon, beryllium (Be), or other suitable material which permits x-rays to pass with a minimum loss of energy. An optimal material for window 26a is diamond, since that material is an excellent heat conductor. In some embodiments, a discrete x-ray transmissive window separate from the target assembly 26 may be used. Using these parameters, the resultant x-rays have sufficient energy to penetrate into soft tissues to a depth of a centimeter or more, the exact depth dependent upon the x-ray energy distribution.

The apparatus of FIG. 1 is particularly adapted for full implantation into a patient, where the housing 12 has a biocompatible outer surface and encloses both a high voltage power supply circuit 12A for establishing a drive voltage for the beam generator 22, and an associated battery 12B for driving that circuit 12A. In this case, an associated controller 12C establishes control of the output voltage of the high power supply circuit 12A, in the manner described below.

The apparatus of FIG. 1 may also be used in a manner where only the probe 14 is inserted into a patient while the housing remains outside the patient, i.e., a transcutaneous form. In the latter form, the various elements shown within housing 12 may alternatively be remotely located.

In the transcutaneous form, the apparatus 10 may be used with an elongated closed end (or cup-shaped) sheath 34, as shown in FIG. 2, having a biocompatible outer surface, for example, fabricated of medical grade aliphatic polyurethane, as manufactured under the trademark Tecoflex® by Thermedics, Inc., Woburn, Mass. With this configuration, the probe 14 is first inserted into the sheath 34. The sheath 34 and probe 14 are then inserted into the patient through the skin. Alternatively, a part may be inserted through the skin and attached to it, as for example a Dermaport® port manufactured by Thermedics Inc., Woburn, Mass. The probe 14 is then inserted into the port.

The lining of the sheath or port can be configured as an x-ray shield by introducing barium sulfate or bismuth trioxide, or other x-ray shielding materials, into the sheath. If necessary, the probe 14 and housing 12 can be secured to the patient's body to prevent any relative motion during the extended time of treatment. An exemplary sheath 34 is shown in FIG. 2.

In one embodiment of the apparatus, the main body of the probe 14 can be made of a magnetically shielding material such as a mu-metal. Alternatively, the probe 14 can be made of a non-magnetic metal, preferably having relatively high values for Young's modulus and elastic limit. Examples of such material include molybdenum,

rhodium or alloys of these materials. The inner or outer surface of probe 14 can then be coated with a high permeability magnetic alloy such as permalloy (approximately 80% nickel and 20% iron), to provide magnetic shielding. The x-ray apparatus 10 can then be used in environments in which there are dc and ac magnetic fields due to electrical power, the field of the earth, or other magnetized bodies nominally capable of deflecting the electron beam from the probe axis.

FIGS. 3A and 3B show an alternative embodiment 10' of the invention adapted for superficial usage, that is for direct placement on the skin of a patient. This form of the invention is particularly useful for x-ray treatment of skin lesions or tumors, or other dermatological applications. In FIGS. 3A and 3B, elements that correspond to elements in the embodiment of FIG. 1 are denoted with the same reference designations. Apparatus 10' generates an electron beam in a channel 40 enclosed within housing 12, where that channel 40 corresponds to probe 14. In the present embodiment, of FIGS. 3A and 3B, the target 26 (elements 26a and 26b) functions as the anode as well as an x-ray emitter. Otherwise, the apparatus 10' is similar to apparatus 10. With the configuration of FIGS. 3A and 3B, low power x-rays may be directed to a desired skin region of a patient.

In all of the above-described embodiments, the x-ray emission element of the target assembly is adapted to be adjacent to or within the region to be irradiated. The proximity of the emission element to the targeted region, e.g. the tumor, eliminates the need for the high voltages of presently used machines, to achieve satisfactory x-ray penetration through the body wall to the tumor site. The low voltage also concentrates the radiation in the targeted tumor, and limits the damage to surrounding tissue and surface skin at the point of penetration. For example, the delivery of 4000 rads, as is required after a mastectomy, with a 40 kV, 20 μ A electron beam, may require approximately 1 to 3 hours of radiation. However, since the x-ray source is, in this preferred embodiment, insertable proximate to, or into, the region-to-be-irradiated risk of incidental radiation exposure to other parts of the patient's body is significantly reduced.

Further, specificity in treating tumors may be achieved by tailoring the target and shield geometry and material at the emission site. This tailoring facilitates the control of energy and the spatial profile of the x-ray emission to ensure homogenous distribution of the radiation throughout the targeted tumor.

FIG. 4 is a schematic representation of the x-ray source apparatus 10 shown in FIG. 1. In that preferred configuration, the housing 12 is divided into a first portion 12' and a second portion 12''. Enclosed within the first housing portion 12' is a rechargeable battery 12B, a recharge network 12D for the battery 12B, which is adapted for use with an external charger 50, and a telemetry network 12E, adapted to be responsive to an external telemetry device 52 to function in the manner described below. That portion 12' is coupled by cables to the second housing portion 12''. The second housing portion 12'' includes the high voltage power supply 12A, controller 12C and the probe 14, as well as the electron beam generating element 22. In the illustrated apparatus 10, the electron beam generator includes a thermionic emitter 22 driven by the power supply 12A. In operation, power supply 12A heats the thermionic emitter 22, which in turn generates electrons which are

then accelerated toward the anode 24. The anode 24 attracts the electrons, but passes them through its central aperture toward the target assembly 26. The controller 12C controls the power supply 12A to dynamically adjust the cathode voltage, the electron beam current, and temporal parameters, or to provide pre-selected voltage, beam current, and temporal parameters.

In the illustrated embodiment, device 52 and network 12E cooperate to permit external control (dynamic or predetermined) control over the power supply 12A and temporal parameters. In embodiments when the housing 12'' is not implanted, but where only probe 14 extends into a patient's body, the controller 12C may directly be used to control operation; in that case there is no need for network 12E.

In an important aspect of the invention, the target assembly 26 may be shaped to emit x-rays in a radiation pattern having a predetermined spatial distribution, and in a predetermined spectral range. This target shaping may be achieved by selecting target materials of known characteristics. For example, as shown in FIGS. 5A and 5B, the emission spectrums for tungsten targets (FIG. 5A) and molybdenum targets (FIG. 5B) are distinct. FIG. 5A shows the x-ray emission spectrum from a tungsten target tube operating at 30 and 50 kV. Note that the bremsstrahlung spectrum predominates, and that x-rays are supplied in a wide energy range. FIG. 5B shows the emission spectrum from a molybdenum target tube, also operating at 30 and 50 kV. Note the near absence of bremsstrahlung x-rays. Note also that the change in tube potential from 30 to 50 kV results in a minor change in the shape of the emission spectrum from a molybdenum target x-ray tube. By selecting the target material, depending upon the type of penetration and the targeted region to be irradiated, the x-ray emission from target assembly 26 may effectively be shaped.

The emission spatial distribution may be also shaped by altering the geometric configuration of target assembly 26. By way of example, the emission element of the target assembly 26 may be shaped such that the electrons directed from the anode will be incident at a predetermined angle or may be selectively directed to different areas of the region from which emission is to occur. In a similar manner, "beam steering" may be used to direct the emitted electron beam to selected surfaces on the emission element, for example, where the target has different emission characteristics in different spatial regions. Control of the electron beam may be achieved under the control of telemetry, or by pre-programming the power source prior to implantation of the apparatus 10.

FIG. 8 shows an exemplary electrostatic beam steering assembly. In the illustrated embodiment, the cathode 22 generates electrons in a manner consistent with the above-described embodiments. The electrons are accelerated through a focussing electrode 23 toward the anode 26, and pass through an aperture 24a toward the target assembly 26. Enroute to target assembly 26, the electrons pass through an electrostatic deflection assembly 30, shown in cross-section at FIG. 8A. The assembly includes four deflectors 32. By varying the voltages applied to the opposing pairs of the deflectors 32, the electrons of the beam entering the assembly along axis 16a are deflected, or "steered" as they travel toward the target assembly 26 along axis 16b. Thus, the beam axis may be controlled to be straight or curved, as desired. As described below, electromagnetic techniques may

alternatively be used to establish beam steering. In the latter case, the electrostatic deflective plates 32 may be replaced with magnetic deflector coils which are driven by currents to establish magnetic fields necessary to achieve a featured beam deflection.

In another form of the beam-steering embodiment, rather than pass through an electrostatic deflection assembly 30, the electron beam passes through a set of magnetic field-generating coils. The coils can be arranged in a configuration similar to the electrostatic deflection plates of the assembly 30. By varying the current through the coils, the resultant magnetic field is produced in a predetermined manner so as to influence the path of the electron beam.

In such a fashion, the electron beam may be steered to hit certain physical locations on a cone-shaped target assembly (FIG. 8), or a target of any other specific geometric configuration. By way of example, in the illustrated embodiment, a beam hitting the angled side of target assembly 26 will result in x-rays emitted off to that side, with little or no incidental radiation to the opposite side of the target assembly.

In another form of the beam-steering embodiment, the x-ray emission characteristics may be controlled by spatially varying the emission parameters (such as radiation peak vs. energy) of the target assembly. By changing the emission peak (as a function of energy) at various points in the target assembly 26, for example, with a "bullseye" spatial pattern, the beam may be steered to regions of relatively high energy x-ray emission, or to regions of relatively low energy x-ray emission. Thus, the beam may be selectively directed to regions of the target assembly to achieve the required x-ray emission characteristic and direction.

In implantable configurations, the power supply 12A and target assembly 26 are preferably enclosed in a metal capsule to prevent current flow from the x-ray source to the patient. The closed housing 12 and probe 14 are, thus, encapsulated in a continuous outer shell of appropriate shielding material, such as those mentioned previously.

The high voltage power supply 12A in each of the illustrated embodiments preferably satisfies three criteria: 1) small in size; 2) high efficiency to enable the use of battery power; and 3) independently variable x-ray tube voltage and current to enable the unit to be programmed for specific applications. A high-frequency, switch-mode power converter is used to meet these requirements. The most appropriate topology for generating low power and high voltage is a flyback voltage converter working in conjunction with a high voltage, Cockroft-Walton-type multiplier. Low-power dissipation, switch-mode power-supply controller-integrated circuits (IC) are currently available for controlling such topologies with few ancillary components.

In order to provide active control of the x-rays, a preferred embodiment of the present invention establishes independent control of cathode voltage and current without the use of a grid electrode. In that form of the invention, an rf ohmic heating current is provided to a thermionic cathode 22, preferably using a transformer-coupled 0.6 volt, 0-300 mA filament power supply floating at the cathode potential of 40 kV. In conventional power supplies for thermionic cathodes, a high voltage isolation transformer would be used to fulfill this function, but such components are quite bulky and thus inappropriate for a small apparatus. In contrast to the prior art, the present invention includes a high volt-

age power supply network which utilizes a dual chain of storage capacitors in the multiplier stage to conduct a radio-frequency (rf) current to the filament while maintaining high voltage direct current (dc) isolation.

The system can be resonated with an inductor (L) at the rf current input, making control of the rf current level possible by changing either the amplitude or the frequency of the drive voltage. This change in the filament current changes the temperature of the filament, thus allowing control of the cathode current emission without changing the cathode-to-anode voltage.

A preferred form of the power supply 12A is shown generally in FIG. 6, and in detailed schematic form in FIG. 7. As shown in FIG. 6, that embodiment includes a flyback switching converter and regulator 280, a 30:1 voltage transformer 282 coupled to a control voltage (or high voltage multiplier input) terminal 282a, and a 10 stage voltage multiplier 284 coupled to a high voltage terminal 22a, and adapted to drive the filament of a thermionic emitter 22. A filament RF power driver and voltage-to-frequency (V/F) converter 290 and an associated RF filament driver 292 are coupled through current control terminal 292a and capacitor C_o by way of a filament drive circuit 286 to the filament of emitter 22.

The multiplier 284 includes a set of $2n$, i.e. 20, series connected diodes and first and second sets of n series coupled capacitors. The set of diodes establishes a unidirectional dc path between the high voltage terminal 22a, through a resistive element R, and to a reference (ground) potential.

The first set of capacitors is coupled between the junction of the first and second diodes (from terminal 22a) and the control voltage terminal 282a. The first $n-1$ capacitors of the first set are each coupled across associated successive pairs of diodes of the set of diodes starting with the second diode (from terminal 22a).

The second set of capacitors is coupled between the high voltage terminal 22a and a reference (ground) potential. The first $n-1$ capacitors of the second set are each coupled across associated successive pairs of diodes of the set of diodes, starting with the first diode (from terminal 284a).

The filament drive circuit 286 includes the voltage multiplier 284 (principally, the second set of capacitors) and a third set of series coupled capacitors. The third set of capacitors is coupled between the thermionic emitter terminal 22b and the current control terminal 292a. Each successive capacitor of the third set is associated with a correspondingly positioned capacitor in the second set, and the capacitor-to-capacitor junctions of the third set are resistively coupled to the correspondingly positioned junctions of the capacitor-to-capacitor junctions of the second set.

A difference amplifier 294 establishes a current feedback loop by driving the RF power driver and V/F converter 290 in response to the detected difference between a current feedback signal on line 295 and an applied emission control signal on line 296. The latter signal may be selectively controlled to establish a desired temporal variation in the x-ray tube cathode current in the filament of emitter (thermionic cathode) 22.

A high voltage amplitude feedback loop is established by the switching converter and regulator 280 in response to the detected difference between a voltage feedback signal on line 297 and an applied high voltage control signal on line 298. The latter signal may be selectively controlled to establish a desired amplitude variation of the potential at the filament of emitter (ther-

mionic cathode) 22. As shown in FIG. 7, the high voltage power supply 12A includes a low power CMOS negative boost converter controller IC 1 which controls the output voltage (at cathode 22) by varying the density of constant energy pulses from the power transformer 282. The output voltage is electrically compared to a programming voltage (from line 298) and, if the output voltage is too low, the frequency of the energy pulses is increased. The control pulse from controller IC 1 is fed to a low power CMOS timer IC 2 where the pulse is shortened and buffered to a level appropriate for driving a high voltage flyback converter (Q1 and T1). High voltage pulses are generated on the secondary of transformer 282 (T1) that are rectified by the n=10 stage diode multiplier 284. Voltage feedback is preferably taken from the first stage of the multiplier 284, since this uses less power and reduces the voltage stress on the feedback resistor R9. The output voltage of the multiplier 284 is well correlated with the voltage at the first stage due to the clamping nature of this type of multiplier.

Cathode emission current control is achieved by sensing the current at the bottom (at R7 and C8) of the n diode chain of multiplier 284, since the only dc path for the cathode current is through the 2n diodes of multiplier 284. This enables sensing the current of the cathode near ground potential instead of at the output potential, which may be 30 kV, for example. This sensed current is converted to a voltage and compared to the emission current programming voltage (signal from line 296) at block 291. If in error, the voltage or frequency of the filament driver is adjusted accordingly in a closed loop manner.

With this configuration, independent control of cathode voltage and current is achieved while maintaining a power efficiency of greater than 70% and a volume of less than 3 cubic inches, meeting the desired design goals of the system.

FIG. 9 shows an exemplary system 300 adapted for x-ray treatment of brain tumors. System 300 includes a stereotactic frame 302 in combination with a low-power x-ray device 10A coupled thereto. In that configuration, x-ray device 10A is generally similar to the x-ray device 10 shown in FIG. 1, but has a cylindrical geometry. Corresponding elements of the two x-ray devices 10 and 10A are identified with the same reference designations. In general, stereotactic frames provide a fixed reference structure relative to the cranium of a patient. While the preferred embodiment described above is particularly adapted for use with this stereotactic frame, other embodiments of the invention might be similarly adapted for use with other frames or with general reference frames, for example, one establishing and operating fixture fixedly referenced to a part of the body other than the head. In the illustrated embodiment of FIG. 9, the stereotactic frame 302 is substantially similar to the Cosman-Roberts-Wells system manufactured by Radionics Inc., Burlington, Mass.

In the illustrated embodiment, the frame 302 establishes a reference XYZ coordinate system disposed about a desired origin point O. The frame 302 includes a generally U-shaped support element 304 defining a reference plane. Four arms 306A, 306B 306C and 306D (not shown) extend out from support frame 304. Each arm has a positioning pin 308. The pins 308 extend generally towards each other from the respective distal tips of arms 306A, 306B, 306C and 306D. In use, the four pins 308 are positioned against a patient's skull to estab-

lish a fixed positional relationship between the frame 302 and the patient's cranium. Thus, the frame 302 defines the reference XYZ coordinate system with respect to the patient's cranium.

An x-ray device support member 310 is coupled to the support element 304 by way of a pair of rotational coupling assemblies 312 and a pair of linear coupling assemblies 314. The x-ray device support member 310 includes an arcuate support track 310A. An x-ray device 10 is coupled to support track 310A by a coupling assembly 316. Coupling assembly 316 provides controlled movement of the x-ray device 10 on a circular path along track 310A and between an inner limit point and an outer limit point along axes (exemplified by axis 316') extending radially inward from the circular path of arcuate track 310A toward the origin point O.

In addition, rotation about the hubs of rotational coupling assemblies 312 allows the x-ray device support member 310 to be rotatably moved about the X axis. The x-ray device support member 310 is translocatable in a direction normal to the plane defined by the X and Y axes (the X-Y plane) by movement along tracks 314A, of the linear coupling assemblies 314. In the illustrative embodiment, a T-groove in tracks 314A mates with a tenon of block 314B which is affixed to member 304, permitting linear motion in the direction perpendicular to the X-Y plane. Set screws 332 in block 314B may be adjusted to lock the x-ray device support member 310 at a set height relative to the support frame 304.

X-ray support member 310 may be moved in the direction of the Z axis by movement of the tenons extending from member 310 in tracks 304A of support element 304. A controlled position of the member 310 along the tracks 304A can be established using locking screws 334.

In addition, support element 304 can be adjustably positioned in the direction of the X axis by sliding member 304 relative to its support member 305, and may be adjustably positioned with three degrees of freedom to establish a desired location of origin point O within the skull of a patient.

The coupling assembly 316 is shown together with an x-ray device 10A, in exploded form, in FIG. 10. As shown, the coupling assembly 316 includes a receiver block 316A, a bushing element 316B, together with complementary shaped portions of the X-ray device 10. As shown, the central axis of probe 14 of x-ray device 10A is coaxial with axis 316'. The electron beam axis 16 of probe 14 is nominally coaxially with axis 316', but may be adjustably varied as described above in conjunction with FIGS. 8 and 8A, and below in conjunction with FIGS. 11 and 12.

The cylindrical bushing element 316B is positioned partially within and coaxially with the receiver block 316A. The bushing element 316B is slidable (in the direction of radial axis 316') and may be selectively locked in place relative to block 316A using a set screw 318A. The bushing element 316B includes a central bore (with diameter D) extending along its central axis.

As noted above, the x-ray device 10A is similar to the x-ray device 10 shown in FIG. 1, but has a generally cylindrically shaped housing 12; the probe 14 includes a cylindrical shoulder portion 14A (having a diameter slightly less than D) immediately adjacent to housing 12, with a main portion with a small diameter (3.0 mm in the preferred embodiment). With this configuration, the x-ray device 10A may be positioned with its axis 16 coaxial with axis 316' and the shoulder portion 14A

slidingly positioned within the bore of bushing element 316B. The relative position of x-ray device 10A may be fixed along axis 316' using set screws 320 of element 316B.

The x-ray device 10A is shown in sectional view in FIG. 11. As shown in FIG. 11, x-ray device 10A includes magnetic deflection subsystem for its electron beam. The deflection subsystem includes magnetic deflection coils 32' positioned about axis 16 within shoulder portion 14A. These coils are driven to adjustably control the position of the beam axis so that the beam is incident on the target of assembly 26 in a desired manner. In the preferred form, radiation generated by device 10A is monitored (for example, by an x-ray detector positioned outside the patient) and the deflector coils are driven accordingly by steering control currents on deflection X1, X2, Y1 and Y2 lines (generated in supply 12') applied to the deflection coils.

In the embodiment of FIG. 10, the microprocessor-based controller is not disposed within the housing 12, but is located external to the housing 12 in a control unit 342. Control unit 342 is coupled to x-ray device 10A by way of cable 342'. The elongated probe 14 of x-ray device 10 is configured so as to allow the probe 14 to pass through the track left by a biopsy needle, thereby permitting easy insertion of the probe 14 into the brain of a patient. For tumors composed of hard tissue, and where a biopsy needle smaller in width than the probe 14 is used, proper penetration into the tumor may require first widening the track left by the biopsy needle with intermediate sized needles.

With this configuration, the tip of probe 14 contains the x-ray emitting target and can be moved in and out relative to the cranial insertion site by movement along the axis 316'. The x-ray device 10A can be secured at a given position along by set screws 318A and 320. The length of probe 14 of x-ray device 10A is chosen such that the tip of probe 14, when fully inserted down to the lower limit point along the axis 316' of 316A, exactly contacts the origin point O; when the x-ray apparatus 10 is fully withdrawn to the upper limit point along axis 316', the distal tip of the probe 14 is intended to be outside the patient's skull. The coordinates of the arcuate support track 310A can be set such that the origin point O is located at the desired epicenter of irradiation. Thus, by the rotation of x-ray device 10A support member 310 and the positioning of the x-ray device 10' along the circumferential track of the arcuate support track 310A and along axis 316', a user can choose the appropriate path (preferably of least destruction) for insertion of probe 14 into a patient's skull, the tip of probe 14 always contacting the origin point O upon full insertion of the probe 14 to the lower limit point.

FIG. 12 shows a schematic diagram of the preferred high voltage power supply 12A for use with the x-ray device 10A of FIGS. 10 and 11. In that power supply, the HV drive signal is a 0 to 9 Volt pulse density modulated drive signal. This signal drives the Flyback Switching FET Q1, which in turn drives the HV Flyback transformer. The HV Flyback transformer steps up the +12 Volts to several thousand volts. The HV multiplier, D1 to D28, in turn steps up the voltage to the desired output voltage of 15 to 40 kV. The voltage FB line provides feedback information to controller 12C, so that the output voltage of the HV multiplier can be held at a constant value.

The Filament + and - lines provide complementary 9 Volt 250 kHz square wave drive signals to FET's Q2

and Q3. These FET's chop the variable Filament DC voltage into an AC voltage, and drive the Filament/HV Isolation Transformer T2. Using a high frequency signal to drive this transformer permits a single turn secondary to drive the x-ray tube filament. This in turn permits miniaturizing the transformer while maintaining the necessary high voltage isolation. The current FB line allows controller 12C to sense the beam current, and the controller then adjusts the Filament DC Voltage for the desired beam current, by providing the appropriate heating current to the thermionic emitter 22. The Deflection X1, X2, Y1, Y2 lines provide current drive signals to the magnetic beam deflection coils.

In the above embodiments, the probe 14 can be coated with a biocompatible outer layer, such as urethane. The outer layer can be applied directly to the body of probe 14, or as a sheath fitting over the probe, such as that illustrated in FIG. 2.

The invention may be embodied in other specific forms without departing from the spirit or essential characteristics thereof. The present embodiments are therefore to be considered in all respects as illustrative and not restrictive, the scope of the invention being indicated by the appended claims rather than by the foregoing description, and all changes which come within the meaning and range of equivalency of the claims are therefore intended to be embraced therein.

What is claimed is:

1. An x-ray system for treating a brain tumor in a patient comprising:

- A. an electron beam activated x-ray source including:
 - i. a housing enclosing an electron beam source having means for generating an electron beam along a beam axis extending out of said housing,
 - ii. an elongated tubular probe extending from said housing about said beam axis, said probe including a target positioned along said beam axis at the end of said probe distal from said housing, said probe having a biocompatible outer surface, and said target being responsive to electrons from said beam incident thereon to generate x-rays,
 - iii. beam positioning means for controlling the position of said beam axis whereby said beam axis intercepts said target,
 - iv. controller means for controlling said means for generating said electron beam and said beam positioning means whereby said generated x-rays are substantially confined to a predetermined volume when said target is positioned within the cranium of said patient,
- B. a reference frame assembly including an arc-shaped frame member extending along a substantially circular circumferential axis, said circumferential axis being characterized by a radius of curvature greater than the largest radius of curvature of the outer surface of the head of said patient, and including means for coupling said frame member to the body of said patient, whereby said frame member is positioned about the head of said patient with a substantially fixed spatial relationship with said head;
- C. coupling assembly for adjustably coupling said x-ray source to said frame member including coupler means for coupling said housing to said frame member in a manner permitting:
 - i. selectively controlled motion of said housing along said circumferential axis of said frame member, and

- ii. selectively controlled motion of said housing along a radial axis, extending from said circumferential axis

whereby said probe may be selectively positioned along said radial axis radially with respect to said circumferential axis from selected points along said circumferential axis.

2. A system according to claim 1 wherein said beam positioning means includes a magnetic deflection subsystem having magnetic deflector coils positioned about said beam and an associated drive circuit for applying a steering control current to said magnetic deflector coils to produce a beam steering magnetic field.

3. A system according to claim 2 further comprising an x-ray detection subsystem including means for detecting selected characteristics of the pattern of x-radiation generated by said electron beam at said target.

4. A system according to claim 3 further comprising means responsive to said detection subsystem for controlling said applied steering control current to said magnetic deflector coils so that said beam is incident on said target.

5. A system according to claim 1 wherein said beam positioning means includes an electric field control assembly including means for compensating for external electric fields in controlling position of said beam axis whereby said beam is incident on said target.

6. A system according to claim 5 wherein said electric field control assembly includes an electrically conductive layer extending along the lateral surface of said tubular probe, and associated means for maintaining said electrically conductive member at a reference potential, thereby establishing a substantially electric field-free region about said beam axis within said tubular probe.

7. A system according to claim 1, further comprising a magnetic field control assembly including means for compensating for external magnetic fields in controlling the position of said beam axis whereby said beam is incident on said target.

8. A system according to claim 7 wherein said magnetic field control assembly includes a high magnetic permeability layer extending along the lateral surface of said tubular probe, thereby establishing a substantially magnetic field-free region about said beam axis within said tubular probe.

9. A system according to claim 1 further comprising electric field control assembly including an electrostatic deflection subsystem within said tubular probe including means for controlling the position of said beam axis, said electric field control assembly including electrostatic deflection plates positioned about said beam axis and an associated drive circuit for selectively applying beam deflecting voltages across said deflection plates.

10. An x-ray system for treating a brain tumor in a patient comprising:

- A. an electron beam activated x-ray source including:
 - i. a housing enclosing an electron beam source having means for generating an electron beam along a beam axis extending out of said housing,
 - ii. an elongated tubular probe extending from said housing about said beam axis, said probe including a target positioned along said beam axis at the end of said probe distal from said housing, said probe having a biocompatible outer surface, and said target being responsive to electrons from said beam incident thereon to generate x-rays,

- iii. beam positioning means for controlling the position of said beam axis whereby said beam axis intercepts said target,

iv. controller means for controlling said means for generating said electron beam and said beam positioning means whereby said generated x-rays are substantially confined to a predetermined volume when said target is positioned within the cranium of said patient,

B. a reference frame assembly including a base member and an arc-shaped frame member tiltably coupled to said base member, said frame member extending along a substantially circular circumferential axis, said circumferential axis being characterized by a radius of curvature greater than the largest radius of curvature of the outer surface of the head of said patient, and including means for coupling said base member to the body of said patient, whereby said base member is positioned about the head of said patient with a substantially fixed spatial relationship with said head;

C. coupling assembly for adjustably coupling said x-ray source to said frame member including coupler means for coupling said housing to said frame member in a manner permitting:

- i. selectively controlled motion of said housing along said circumferential axis of said frame member, and
- ii. selectively controlled motion of said housing along a radial axis, extending from said circumferential axis

whereby said probe may be selectively positioned along said radial axis radially with respect to said circumferential axis from selected points along said circumferential axis.

11. A system according to claim 10 wherein said beam positioning means includes a magnetic deflection subsystem having magnetic deflector coils positioned about said beam and an associated drive circuit for applying a steering control current to said magnetic deflector coils to produce a beam steering magnetic field.

12. A system according to claim 11 further comprising an x-ray detection subsystem including means for detecting selected characteristics of the pattern of x-radiation generated by said electron beam at said target.

13. A system according to claim 12 further comprising means responsive to said detection subsystem for controlling said applied steering control current to said magnetic deflector coils so that said beam is incident on said target.

14. A system according to claim 10 wherein said beam positioning means includes an electric field control assembly including means for compensating for external electric fields in controlling position of said beam axis whereby said beam is incident on said target.

15. A system according to claim 14 wherein said electric field control assembly includes an electrically conductive layer extending along the lateral surface of said tubular probe, and associated means for maintaining said electrically conductive member at a reference potential, thereby establishing a substantially electric field-free region about said beam axis within said tubular probe.

16. A system according to claim 10, further comprising a magnetic field control assembly including means for compensating for external magnetic fields in con-

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trolling the position of said beam axis whereby said beam is incident on said target.

17. A system according to claim 16 wherein said magnetic field control assembly includes a high mag-
netic permeability layer extending along the lateral
surface of said tubular probe, thereby establishing a
substantially magnetic field-free region about said beam
axis within said tubular probe.

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18. A system according to claim 10 further compris-
ing electric field control assembly including an electro-
static deflection subsystem within said tubular probe
including means for controlling the position of said
beam axis, said electric field control assembly including
electrostatic deflection plates positioned about said
beam axis and an associated drive circuit for selectively
applying beam deflecting voltages across said deflection
plates.

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